Ultrafast MR Imaging

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Motivation

Magnetic resonance imaging (MRI) is unique in its great flexibility and wide range of applications. Not only can it provide high resolution anatomical images, but it can also provide information about physiological processes that is unobtainable from other modalities. However, because signal generation in MRI depends on the intrinsic properties of the sample (or subject) being studied, the achievable signal-to-noise ratio (SNR) is strongly dependent on the acquisition time. As a result, longer scan times are often necessary to produce high quality results.

One of the principle motivating factors in ultrafast imaging is the desire to reduce this necessary scan time as much as possible. There are several obvious reasons why a faster acquisition is advantageous. In terms of patient comfort, for very ill patients that are unable to tolerate long periods in the magnet a faster acquisition can be the difference between a successful and unsuccessful result. In addition, the clinician might be interested in imaging a physiological process with a time scale that is short compared to traditional MR scan times. Examples range from cardiac dynamics, with typical R-R intervals on the order of 1 sec (implying the need to capture images with resolutions on the order of tens of milliseconds), to the desire to eliminate respiration artifacts via a breath-hold (which might be on the order of 15 seconds), to the desire to image contrast passage (which can take anywhere from 10-40 seconds depending on transit time).

There is of course a limit to how fast a scan can be performed. The role of the MR acquisition is to acquire a set of k-space data that is used to form a set of images [1]. Ultimately, a minimum amount of k-space data must be collected (which implies a minimum scan time) in order to produce enough resolution in the final image to render it diagnostically useful. Furthermore the image SNR is proportional to the voxel size and the time spent acquiring data [2]:

$$SNR \propto \Delta V \times \sqrt{T_{acq}}$$

Therefore the SNR in the final image will decrease as less time is spent acquiring data. Some of this loss can be made up with the use of better coil arrays or with contrast agents, but ultimately SNR considerations will limit scan time reductions. We will also see that fast scanning techniques have a greater sensitivity to artifacts which can make them unsuitable for certain applications.

Echo Planar Imaging (EPI)

To introduce the concepts involved in fast imaging, we will begin with echo-planar imaging. EPI was first described in 1977 by Sir Peter Mansfield at the University of Nottingham [3], and with the evolution of better and more sophisticated MR imaging systems the original technique has evolved considerably. Of the several variations in EPI we will focus on the simplest implementation, that of a single-shot, gradient-echo (GRE) acquisition [4].

To understand how EPI works, it helps to compare it with standard fast spin echo (FSE) imaging. In an FSE acquisition, a single 90° RF excitation is used to create transverse magnetization which is then refocused with several 180° pulses over the course of the sequence repetition time (TR). After each 180° pulse a line of k-space is acquired, and the total number of lines acquired per TR is determined by the echo train length (ETL). If we assume a TR of 1 s, an ETL of 10, and that we wish to acquire 200 total lines of k-space, our total imaging time will be (200/10) × 1 s = 20 s.

In GRE-EPI, each TR also begins with a 90° RF excitation pulse to create transverse magnetization. But unlike FSE, multiple k-space lines are collected by oscillating the readout gradient continuously to refocus the transverse magnetization into a successive series of echoes. As each echo is acquired, the phase encode gradient is increased slightly, or "blipped", in order to move to the next line of k-space [5]. In a single-shot GRE-EPI acquisition all the needed lines of k-space for a complete image are acquired in a single TR. The minimum TR is dependent on the image resolution and the number of slices desired, but a single slice low resolution (64 by 64 pixels) image can be completely acquired in as little as 40 ms.

Clearly EPI provides a significant reduction in imaging time. However this time savings often comes at the cost of a reduction in image quality and an increase in the number of artifacts. One of the greatest disadvantages to single shot GRE-EPI is the limitation on the achievable image resolution. Because each echo is formed through gradient refocusing, the transverse signal evolution is dominated by $T2^*$ decay. And because $T2^*$ time constants tend to be quite short, image matrix sizes greater than 128 × 128 are impractical because of the rapid decrease in transverse magnetization [6]. In addition, the falloff in signal from $T2^*$ decay will modulate the acquired k-space data in such a way that the resulting image will suffer from increased blurring [7].

A characteristic of many fast imaging sequences is their long readouts. While this added acquisition time helps increase the efficiency of the sequence, it also tends to make the sequence more susceptible to off-resonance artifacts. A common problem is the signal from lipids, which at 1.5 T resonate at a frequency difference of about 220 Hz from water. During the long EPI readout, these off-resonant spins will produce phase errors in the phase-encode direction of the sequence. As a result, any lipids that are present in the imaging volume will be shifted in the phase-encode direction, often by a significant fraction of the field-of-view. This problem is resolved with the use of special RF pulses that are designed to excite only water [8]. In addition to these chemical shift artifacts, susceptibility differences between adjacent regions of different structures will also produce artifacts in EPI images. These artifacts are most common at air-tissue interfaces (such as at the border of the sinus cavities), and will appear to geometrically distort the local anatomy [9].

All of these artifacts can be reduced by decreasing the amount of time spent during readout. One of the most successful ways of achieving this has been through the development of MR imaging technology. MRI systems have gone from gradient limits of 10 mT/m and slew rates of 17 mT/m/ms to clinical systems today that have typical configurations of 40 mT/m gradient limits and slew rates of 230 mT/m/ms. The result has been a decrease in readout duration with a corresponding decrease in artifacts. However, the increase in gradient slew rates can lead to peripheral nerve stimulation (PNS) in subjects which is potentially dangerous at high levels [10]. All commercially

available imaging sequences are designed to operate below the PNS threshold. Nonetheless, some subjects can report physical sensations such as tingling or twitching during rapid imaging studies.

Multi-shot EPI

While single shot EPI sequences provide great reductions in scan time, the long readouts that make this possible limit the achievable resolution and lead to an increase in artifacts. If slightly longer scan times are permissible, multi-shot techniques can be used to address these limitations. The concept behind a multi-shot implementation is very simple: instead of covering all of k-space in one excitation, the complete k-space data set is acquired over several TRs. As an example, consider a single shot technique that acquires 128 phase encodes with a readout of 100 ms and a TR of 1 s. A 4-shot implementation would acquire 32 phase encodes in a single TR using a readout time of just 25 ms, and the complete set of data would be acquired over a period of 4 TRs. Provided that the TR length is kept constant, the scan time would increase by a factor of four. But the advantage of this approach is that there is less time in the shorter readout over which phase errors can accrue. As a result, multi-shot techniques are less affected by chemical shift and susceptibility artifacts. And because there is less time during data acquisition for $T2^*$ decay, multi-shot images will contain less blurring than single-shot techniques. Higher resolution images can then be acquired by adding more TRs as necessary at the expense of an increase in scan time.

Spiral Imaging

A second type of fast imaging sequence is the spiral sequence. The first spiral images were published by Ahn in 1986 and then further developed by Meyer in 1992 for coronary imaging [11, 12]. Spiral sequences use oscillating waveforms on both in-plane axes during readout. As a result, the beginning of the data acquisition starts at the origin of k-space and spirals outward, which is how the technique gets its name. In using both in-plane gradients equally the spiral sequence is more efficient from a hardware point of view than an EPI sequence. Because of its geometry the spiral trajectory is capable of sampling a given region of k-space more efficiently than a Cartesian acquisition, and as a result shorter readout times are necessary. Finally, the oscillating gradients are inherently moment compensated, which means that flow-induced phase errors will be periodically refocused.

Like any rapid imaging technique with long readouts, spiral imaging is more susceptible to image artifacts than standard imaging techniques. The presence of off-resonant spins such as lipids will result in blurring throughout the image (compared with the shift of the fat signal in EPI). Therefore spiral acquisitions generally also use spatial-spectral pulses to eliminate the signal from these unwanted spins.

Spiral imaging has tended to be less widely adopted than acquisitions that use traditional Cartesian sampling. Part of this reason is that because the k-space data are not acquired on a rectangular grid the spiral reconstruction is more complicated. However this has become less of an issue with the development of more sophisticated reconstruction algorithms. In addition, partial Fourier and rectangular FOV acquisitions are difficult to implement because of the symmetry of the gradient activity on both in-plane axes. Imaging artifacts due to susceptibility can be considered in some cases to be

unacceptable [13]. However, spiral imaging has been shown to have some advantages over EPI in functional magnetic resonance (fMRI) studies due to its lower sensitivity to brain motion and it's more efficient acquisition [14].

Parallel Imaging

Parallel Imaging (PI) is a relatively new development in MRI that speeds up image acquisition by using the spatial information from each element of multiple coil arrays to replace data that would normally be acquired through phase encoding [15]. The implementation of a PI acquisition involves two steps. In the first step the desired imaging sequence simply acquires some fraction of the number of phase encodes that would normally be acquired. It is important to note here that PI is not restricted to a particular type of sequence, as any type of acquisition can be reconfigured to acquire less data. And because less data are acquired, the imaging time is reduced by a corresponding amount. The problem of course is that because only a fraction of the needed phase-encodes have been acquired the resulting images will be severely aliased unless these missing data are somehow replaced.

This is done using multiple coil arrays, the use of which forms the second part of the PI acquisition strategy. The idea is that because each coil element has a different spatially dependent RF sensitivity profile, this information can be used to supplement the gradient localization done through conventional phase encoding. Currently there are several different reconstruction methods which synthesize the missing data during the reconstruction. However all of these methods fall into two general categories. The first encompasses reconstructions that attempt to fill in the missing k-space data before the Fourier transformation. A well-known example of this approach is the GeneRalized Auto-calibrating Partially Parallel Acquisition (GRAPPA) [16]. The second category involves reconstructions that first perform the Fourier transformation and then attempt to remove the aliased signal in the image domain. The most well-known of these techniques is SENSitivity-Encoded MRI (SENSE) [17].

The clear advantage with PI techniques is the reduction in scan time. One is theoretically able to reduce the scan time by a factor of anywhere up to the number of coils in the array (*e.g.*, for a 4-element coil array reduction factors from 1 to 4 are possible). Image quality typically suffers at the extreme end of this range, and so more typical reduction factors would be 2 to 3 (for four and eight-channel coils, respectively). Another advantage arises when PI is used in EPI sequences. Reducing the number of acquired phase encodes results in readout lengths that are shortened by the reduction factor. The results are similar to those gained in multi-shot EPI: the shorter readouts leave less time for phase errors to accrue, leading to fewer artifacts from chemical shift and susceptibility effects.

The most obvious disadvantage is the reduction in the image SNR. As the scan time is lowered the subsequent image SNR is reduced by the square root of the reduction factor. In actuality the SNR is reduced slightly more due to a so-called geometry factor which reflects imperfections in the coil coverage. This has the added implication that the SNR in the resulting image will be spatially dependent. Using coil arrays specifically designed for PI will help mitigate these effects. All PI reconstruction techniques require a low resolution k-space data set that is used to calculate the RF sensitivity profile for each coil. These data can either be acquired with a separate calibration scan prior to the

PI sequence or as extra data during the PI sequence itself. In the latter case some of the scan time reduction advantage is lost, but this is not usually a significant problem.

Parallel imaging is currently a very active area of research in MR. As manufactures continue to increase support for multiple coil arrays PI techniques will find their way into virtually every application. Furthermore, the added SNR that comes from imaging at higher fields means that PI will be even more important for systems at 3T and above [18].

Conclusions

Ultrafast imaging techniques come in a wide variety of implementations, but in every case the goal is rapid data acquisition that can provide the temporal resolution needed to investigate the dynamics of different physiological processes. EPI sequences have become the foundation for many clinical neurological applications such as diffusion and perfusion imaging, as well as the basis for much of the work being presently done in functional MRI (fMRI). Spiral imaging is an attractive alternative to EPI and has become well established in fMRI studies as well. Both types of sequences provide significant time savings over conventional techniques such as spin-echo or fast spin-echo imaging. The tradeoff comes in the form of increased sensitivity to artifacts from chemical shift effects (*e.g.*, signal from fat) and susceptibility effects (*e.g.*, transitions from tissue to air space). With careful protocol design many of these issues can be minimized, resulting in the ability to increase coverage with improved temporal resolution.

Parallel imaging is a relatively new form of fast imaging that refers not to a specific type of sequence, but instead to an approach for acquiring and reconstructing data more quickly from any type of sequence. The key to PI techniques lies in the use of multiple RF coils to provide information regarding spatial encoding that then takes the place of some of the traditional phase encoding. In this fashion any sequence can be run in a fraction of its usual time, implying that the scan time for virtually any application can be decreased with a PI acquisition. This is currently a very active area of imaging research that is driving many of the technological advances in imaging hardware. The future will see the development of systems with the capacity for greater numbers of coils in order to take advantages of these developments.

References

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